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The effect of control strategies for an active back-support exoskeleton on spine loading and kinematics during lifting

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ABSTRACT

With mechanical loading as the main risk factor for LBP, exoskeletons (EXO) are designed to reduce the load on the back by taking over part of the moment normally generated by back muscles. The present study investigated the effect of an active exoskeleton, controlled using three different control modes (INCLINATION, EMG & HYBRID), on spinal compression forces during lifting with various techniques.

Ten healthy male subjects lifted a 15 kg box, with three lifting techniques (free, squat & stoop), each of which was performed four times, once without EXO and once each with the three different control modes. Using inverse dynamics, we calculated L5/S1 joint moments. Subsequently, we estimated spine forces using an EMG-assisted trunk model.

Peak compression forces substantially decreased by 17.8% when wearing the EXO compared to NO EXO. However, this reduction was partly, by about one third, attributable to a reduction of 25% in peak lifting speed when wearing the EXO. While subtle differences in back load patterns were seen between the three control modes, no differences in peak compression forces were found. In part, this may be related to limitations in the torque generating capacity of the EXO. Therefore, with the current limitations of the motors it was impossible to determine which of the control modes was best. Despite these limitations, the EXO still reduced both peak and cumulative compression forces by about 18%.

1. Introduction

Low-back pain (LBP) is a major and still growing problem world-wide (Hartvigsen et al., 2018; Vos et al., 2016), with a lifetime prevalence between 75 and 84% (Thiese et al., 2014). Mechanical loading of the low back has been shown to be an important risk factor for the development of LBP (Coenen et al., 2014; Coenen et al., 2013; da Costa and Vieira, 2010; Griffith et al., 2012; Kuiper et al., 2005; Norman et al., 1998).

Although physically demanding jobs have become less prevalent due to mechanization and automation, some remain because they require the flexibility of the human. In these jobs, back loading can be reduced, by up to 20%, with ergonomic interventions (e.g. adjusting lifting technique, foot positioning, lifting height etc.) (Burgess-Limerick and Abernethy, 1998; Hoozemans et al., 2008; Kingma et al., 2004, 2016; Marras et al., 1999). However, these ergonomic interventions are not always applicable, so that other solutions are still needed.

More recently, body-worn assistive devices (exoskeletons) have been developed. The main aim of industrial use exoskeletons (EXO) is to prevent injury, while preserving the versatility of the human in the production process. Specifically, low-back exoskeletons are designed to reduce the load on the back by taking over part of the moment normally generated by back muscles to counteract moments due to gravity and inertia. A basic distinction is made between passive and active exoskeletons (de Looze et al., 2016). In the former, support patterns are determined ahead of time as part of the mechanical design, e.g., by spring deformation induced by angular changes of joints. In the latter, powered actuators are used to generate supportive torques and are completely controlled by the corresponding assistive strategy. With information from the environment and the user, captured by sensors, a program (control mode) converts the information to commands for the actuators.

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There are some studies that implemented different type of control strategies, based on segment angles, interaction forces, muscle forces and combinations of these variables (Fleischer and Hommel, 2008; Giovacchini et al., 2015; Grazi et al., 2018; Hara and Sankai, 2010; Hayashi et al., 2005; Saccares et al., 2017). However, none of these studies used L5/S1 compression forces as outcome measure and some only quantified the effects via questionnaires.

The present study investigated the effect of an active EXO, controlled using three different control modes, on L5/S1 moments, muscle activation, spine kinematics, and L5/S1 spinal compression forces during lifting with various techniques. The control modes were INCLINATION (based on trunk inclination), EMG (based on forearm EMG) & HYBRID (combining INCLINATION and EMG). We hypothesized that HYBRID would lead to the largest reduction of spine loads over the whole lifting cycle, because it both supports loading due to trunk bending and the additional loading due to lifting the box. However, no major differences in peak loading were expected, as all three control modes were designed to generate maximal support around the moment of peak loading.

2. Methods

2.1. Exoskeleton

The device was developed as part of the EU-funded project Robo-Mate, a revised second version (Mk2B) was used in this experiment, see Fig. 1. In total, the EXO has a mass of 11 kg. Details of the device (EXO) can be found in Toxiri et al. (2018). In short, the EXO spans the trunk and upper legs, with a waist/abdomen fixation. Two actuators, approximately aligned with the hip flexion-extension axis, can generate a maximum torque of 20 Nm each. The torques are approximately limited to the sagittal plane. With the actuators being active, force is applied at the shoulders and the upper legs generating an extension moment in the same direction as the back-muscle extension moment. The torques produced by the actuators were controlled using three different control strategies (Fig. 2). In INCLINATION mode, the torques applied were scaled with the sine of the trunk inclination angle (with respect to the vertical), with the maximum torque at 90 degrees of trunk inclination. In EMG mode, support torques were applied when loads were lifted. With a commercial bracelet (Thalmic Labs Inc., Kitchener ON, Canada), placed just below the elbow, forearm EMG was measured to determine instants when holding and lifting an object. In a trial in which the participant held a box of 15 kg in front of the body, the rectified and 3 Hz low pass filtered EMG signal of the forearms was set as a reference. Torques were then linearly scaled to this reference value, so that approximately 2×20 Nm was applied at times when the participants were fully carrying the weight of the box. In HYBRID mode, the two previous control modes were added together. However, each of the control modes could provide at most half of the maximum of 40 Nm, so that when both control modes were fully active, still 40 Nm was applied.

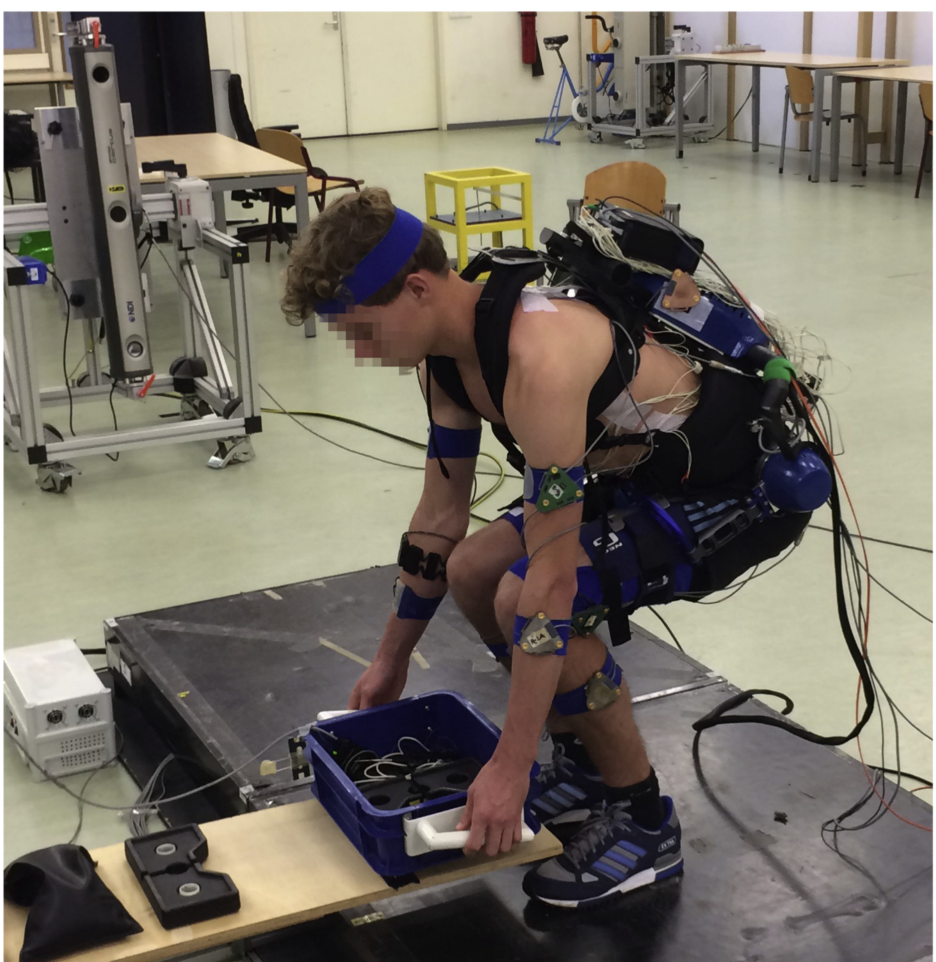

Fig. 1. Picture of a subject lifting a load while using the EXO.

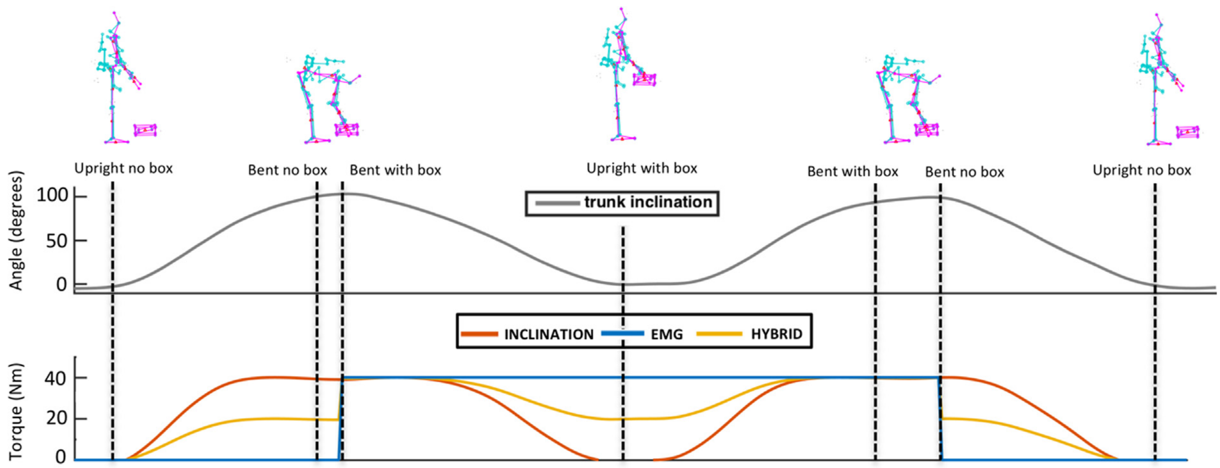

Fig. 2. Simplified illustration of the implemented control strategies. INCLINATION (red) follows the sine of torso inclination regardless of whether the user is holding the object. The EMG mode (blue) switches on when the user holds the object. HYBRID (yellow) is a combination of the two behaviors, in which each component contributes half of the reference torque. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2.2. Subjects and experimental procedures

Ten healthy male subjects (25.0 ± 6.9 years, 70.9 ± 8.8 kg, 1.77 ± 0.06 m), participated in the study, which was approved by the local ethics committee. None of the participants had a history of low-back pain. After providing written informed consent, subjects performed maximum voluntary contractions for both the back and abdominal muscles. Next, anthropometric data were obtained and participants were fitted and familiarized with the EXO. After all instrumentation was placed on the subjects and calibration measurements were performed, participants were instructed to complete a lifting task with three different techniques; FREE, SQUAT ('bend the knees and try to keep the back as straight as possible') and STOOP ('bend from the back and keep the knees as straight as possible'), once with NO EXO (no EXO is worn) and three times with the EXO (INCLINATION, EMG & HYBRID). Prior to the data collection in each of the control modes, participants were familiarized with the current control mode by freely moving and bending the trunk, and then to use each technique with the three EXO control modes. Each lift was replicated three times in each combination of technique and EXO. Participants had to grasp a 15 kg box (dimensions: width × height × depth = 35 × 10 × 25 cm) with handles located at 10 cm above ankle height, return to upright stance (box lifting), and subsequently place the box back and return to upright stance once more (box lowering). The order of the EXO control conditions was randomized over subjects to minimize potential order-related confounding effects, but always started with the NO EXO condition. This was done to limit the number of times the EXO had to be taken off or put on, so that the risk of displacing markers was as low as possible. Subjects always started with the FREE lift to limit the interference of the instructed lifts on the FREE lifting technique in each of the EXO conditions. The order of the SQUAT and STOOP lifts was randomized in each of the EXO control conditions. Sufficient rest between tasks was ensured by having 5 min rest between control conditions.

2.3. Instrumentation and data pre-processing

A single custom-made 1.0 × 1.0 m force plate was used to measure ground reaction forces at 200 samples/s. Kinematics were collected at a sample rate of 50 samples/s using an opto-electronic 3D movement registration system (Certus, Optotrak, Norton Digital Inc.). LED cluster markers were attached to body segments (feet with lower legs (modeled as one segment), upper legs, pelvis, trunk, head, upper arms and forearms with hands) and related to anatomical landmarks using pointer measurements (Cappozzo et al., 1995). Shape of the structures to which the pelvis and trunk cluster markers were attached, were adapted relative to previous work (Faber et al., 2011), to avoid interference with the EXO. In addition, two cluster markers were attached to the pelvic and trunk parts of the exoskeleton. Within the EXO, inclination of the trunk part was measured using an inertial measurement unit (IMU) and the hip EXO joints' angles were measured using an encoder. Torques within the actuators were measured using a strain gauge-based torque sensor.

Ten pairs of surface EMG electrodes were attached to the trunk muscles (Rectus Abdominis, External Oblique, Internal Oblique, Iliocostalis (IL), and Longissimus lumborum (LL); see Kingma et al. (2010)) after abrasion and cleaning with alcohol. EMG data were amplified (Porti-17TM, TMS, Enschede, The Netherlands), band-pass filtered (10–400 Hz) and A–D converted (22 bits at 1000 Hz) and stored synchronized to Optotrak and force plate data. Data from the exoskeleton were synchronized using cross correlation of the EXO hip joint angles measured with Optotrak and by the encoder within the EXO.

2.4. Data analysis

Data were low-pass filtered using a bi-directional 2nd order Butterworth filter at a cut-off frequency of 5 Hz for the marker data and 10 Hz for the force plate data. Total L5/S1 flexion-extension moments (M_{L5S1_total}), generated by subject plus EXO were calculated based on the GRF and kinematics, using a bottom-up inverse dynamics model (Kingma et al., 1996). The moment generated by the subject ($M_{L5S1\_subject}$) was calculated by subtracting the moment generated by the device (M_{L5S1_EXO}) from M_{L5S1_total}. Off-line, EMG signals were full-wave rectified and low-pass filtered at 2.5 Hz (Potvin et al., 1996). EMG data were normalized to maximum voluntary contractions (McGill, 1991) and used as input to an EMG driven muscle model. The model has been described in more detail previously (van Dieen, 1997; van Dieen and Kingma, 2005), and consisted of 90 muscle slips (McGill, 1996; Stokes and Gardner-Morse, 1995), with wrapping points at L4 and T12. Muscle forces were estimated as the product of the assumed maximum muscle stress, normalized EMG amplitude and correction factors for the instantaneous muscle length (Woittiez et al., 1984) and contraction velocity (van Zandwijk, 1998). For each participant, a best fit between net moments and muscle moments was obtained by constrained optimization, (MATLAB optimization toolbox, The Mathworks Inc. Natick, MA), over all lifts performed in the NO EXO condition by a participant. Three parameters for each participant were optimized: the gain, i.e. a scaling factor between EMG amplitude and muscle stress, the position of the passive length-tension curve relative to the muscle optimum length, and a scaling factor for the passive length-tension curve. The optimized values were also used in the three EXO control conditions, without optimizing them again. Finally, compression forces at the L5/S1 intervertebral joint were obtained by summing muscle forces and net reaction forces in the L5/S1 axis system. Lumbar angles were obtained by Euler decomposition of thorax relative to the pelvic anatomical axes (order: flexion-extension, lateral bending, axial rotation).

Mean timeseries for the statistical analysis were derived by aligning the three repetitions using cross-correlations based on the trunk flexion angle. Subsequently the average was taken over the three repetitions. Next, signals were cut (to a box lifting cycle and a box lowering cycle separately) and time normalized to be able to average over subjects with different lifting speeds.

2.5. Statistics

Peak compression forces, peak M_{L5S1_total}, peak $M_{L5S1\_subject}$, averaged IL and LL muscle activity (net lumbar extensor activity), averaged RA and EO muscle activity (net abdominal activity), peak lumbar flexion angle and trunk angular velocity were tested using separate two-way repeated measures ANOVAs with the EXO conditions (NO EXO, INCLINATION, EMG & HYBRID), technique (FREE, SQUAT & STOOP) and their interaction. When a main effect of EXO conditions or an interaction between EXO conditions and technique was present, effects were further explored using Bonferroni post-hoc tests. To examine effects of the EXO conditions on resulting compression force patterns in more detail, compression forces were statistically tested along the complete time series using one-dimensional statistical parametric mapping (SPM1D) (Pataky et al., 2013). SPM1D one-way repeated measures ANOVAs were conducted two times, once with the factor EXO conditions with four levels (NO EXO, INCLINATION, EMG & HYBRID) and once with the factor control with three levels (INCLINATION, EMG & HYBRID). Tests were conducted, for each of the techniques separately (FREE, SQUAT & STOOP) and for box lifting and box lowering separately. A significance level of $p < 0.05$ was used.

3. Results

Correlations (R^2) of the fit between $M_{L5S1\_subject}$ and the EMG driven model moment ranged from 0.79 to 0.89, mean squared differences ranged from 12.5 to 24.0 Nm (5–10% of the highest average peak moment) over subjects (Fig. 3).

3.1. Effects on peak variables

In line with our hypothesis, we found no differences between peak compression forces of the three different EXO control modes, but a substantial reduction in compression force was found compared to NO EXO. An overview of statistical results can be found in Table 1.

Peak compression with EXO was, for all control modes and lifting techniques, significantly lower compared to NO EXO (on average 17.8%, ranging from 14% to 22% over control modes and lifting techniques), except for the EMG mode during stoop lifting (Fig. 4). In line, peak $M_{L5S1\_subject}$ was always significantly lower when using the EXO compared to NO EXO. However, this change was not only due to the EXO support which is the difference between the total moment and subject moment for the EXO conditions (on average 16 Nm), but also to a change in lifting behavior, reflected in the difference between the total moment in the EXO conditions and the total moment in the NO EXO condition (on average 12 Nm). Peak trunk angular velocity was reduced by about 25%. in all EXO conditions relative to NO EXO. In line with the reduction in $M_{L5S1\_subject}$, a substantial reduction in lumbar EMG (by on average 13.3% MVC) was found in all EXO conditions compared to NO EXO. In addition, peak lumbar flexion was reduced, by on average 15.9°, in all EXO conditions compared to NO EXO. No differences between the three EXO control modes were found except for a non-substantially (<3%), but significantly, lower peak

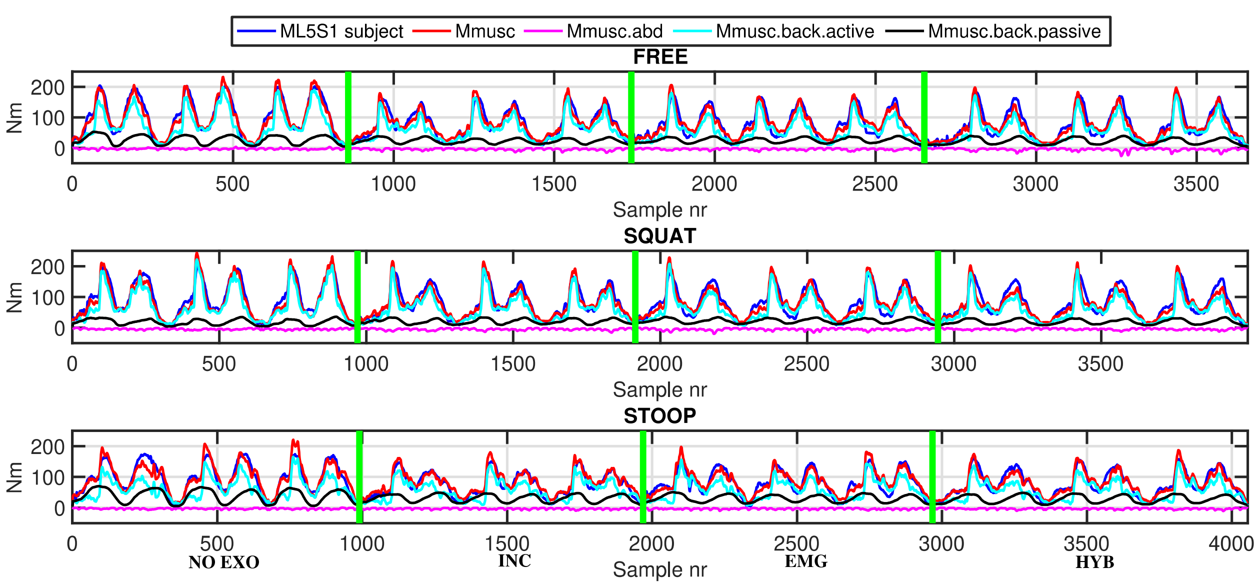

Fig. 3. Time series of the net moments from inverse dynamics, from the EMG driven muscle model, moments generated by the abdominals and the active and passive moments generated by the back muscles for one representative participant. The time series are shown for all four conditions, indicated by the vertical green lines and the three lifting techniques. Note that, the optimization was performed on the NO EXO data only. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Table 1
P-values of repeated measures ANOVA with control condition (NO EXO, INCLINATION, EMG & HYBRID), lifting technique and their interaction (FREE, SQUAT & STOOP) as independent variables. Pairwise comparisons were only performed for variables with a significant main effect of control, and only to compare control conditions. Significant ($p < 0.05$) results are indicated in bold. For variables with both main effects of control and interaction of control with technique, these comparisons were performed per technique, and are displayed in Fig. 3.

	Main effect	Main effect	Interaction	Pairwise comparisons control					
	Control	Technique	Control * Technique	1–2		1–3		1–4	
	p	*p*	*p*	Diff	*p*	Diff	*p*	Diff	*p*
Compression	**<0.001**	0.927	**0.049**						
M_{L5S1_total}	0.267	**<0.001**	0.868						
IL + LL activity	**<0.001**	0.054	**0.045**						
$M_{L5S1\_subject}$	**<0.001**	**<0.001**	0.689	**29.0**	**0.001**	**26.4**	**<0.001**	**27.6**	**<0.001**
Lumbar flexion	**<0.001**	**<0.001**	**<0.001**						
Trunk angle velocity	**<0.001**	**<0.001**	**0.001**						
RA + EO activity	0.696	0.209	0.810						

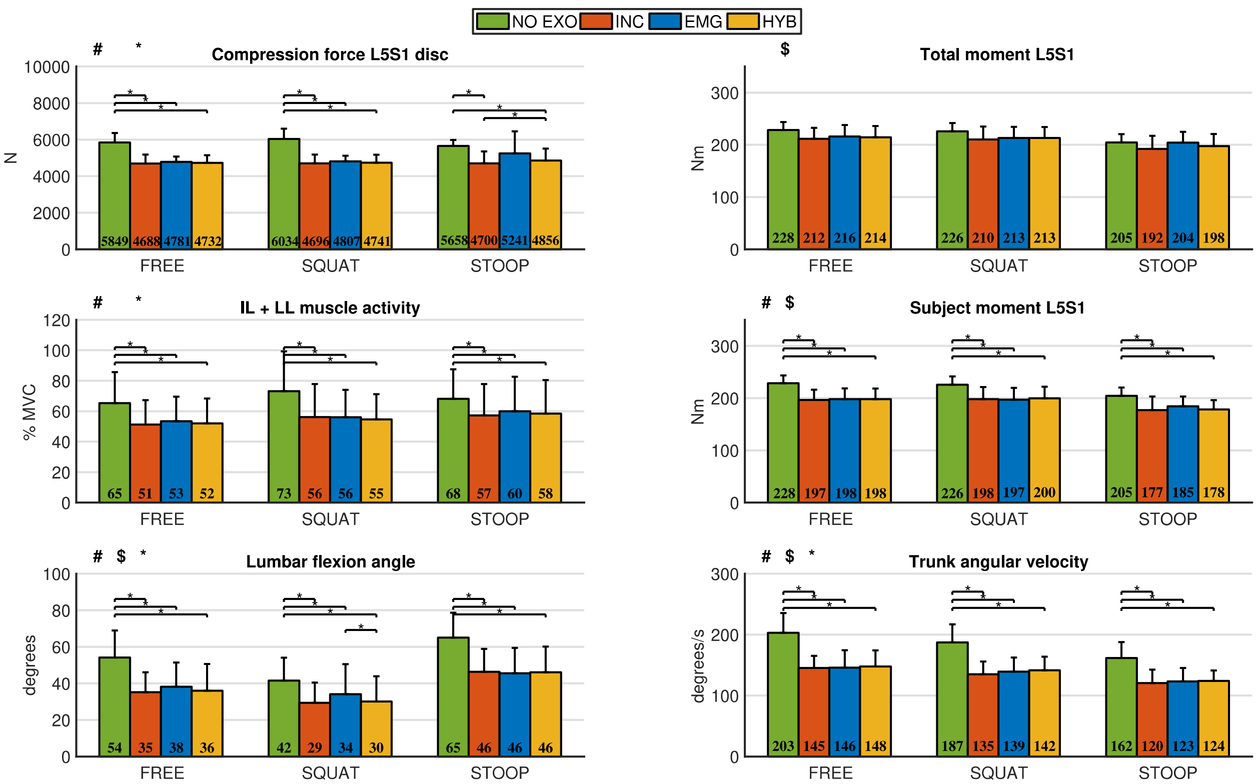

Fig. 4. Peak outcome variables with peak compression force, back muscle activity and lumbar flexion in the first column and M_{L5S1_total}, $M_{L5S1\_subject}$, and trunk angular velocity in the second column. Error bars indicate SD. A main effect of control is indicated with a #. A main effect of technique is indicated with a $. * indicates a significant interaction between control and technique. Horizontal bars indicated a significant post-hoc differences between conditions.

compression force with INCLINATION compared to HYBRID during STOOP lifting.

3.2. Time series analysis, comparing EXO conditions

Not only at the peaks, but also during box lifting and box lowering cycle, compression forces were significantly lower compared to NO EXO during phases of forward bending (Fig. 5).

Compression forces did significantly differ between the three control modes at some instants during the FREE, SQUAT and STOOP techniques. In the box lifting cycle, during FREE lifting, the significant phase was related to bending forward without box (20–40% of the cycle). In this phase, INCLINATION and HYBRID modes provided on average around 20–25 Nm and 15–20 Nm of support, respectively, against about 5 Nm for EMG control (Fig. 6), which explains the lower compression forces with INCLINATION and HYBRID modes compared to EMG mode. The same pattern is visible for the other lifting techniques, although during the SQUAT technique, differences between compression forces were less clear due to the smaller support given in INCLINATION mode due to smaller inclination angles compared to FREE and STOOP. In box lowering, significant control mode effects were found only for FREE and SQUAT lifting when standing upright while holding the load. In this phase, as expected, compression forces with EMG were lowest, followed by HYBRID and INCLINATION. Support of the EXO was about 30, 18 and 8 Nm respectively for EMG, HYBRID and INCLINATION during the significant phases, for both FREE and SQUAT lifting.

No differences were found in peak compression forces between the three control modes. This may be related to limitations in the power generating capacity of the EXO. In fact, the actual torques generated by the EXO, during periods with negative angular accel-

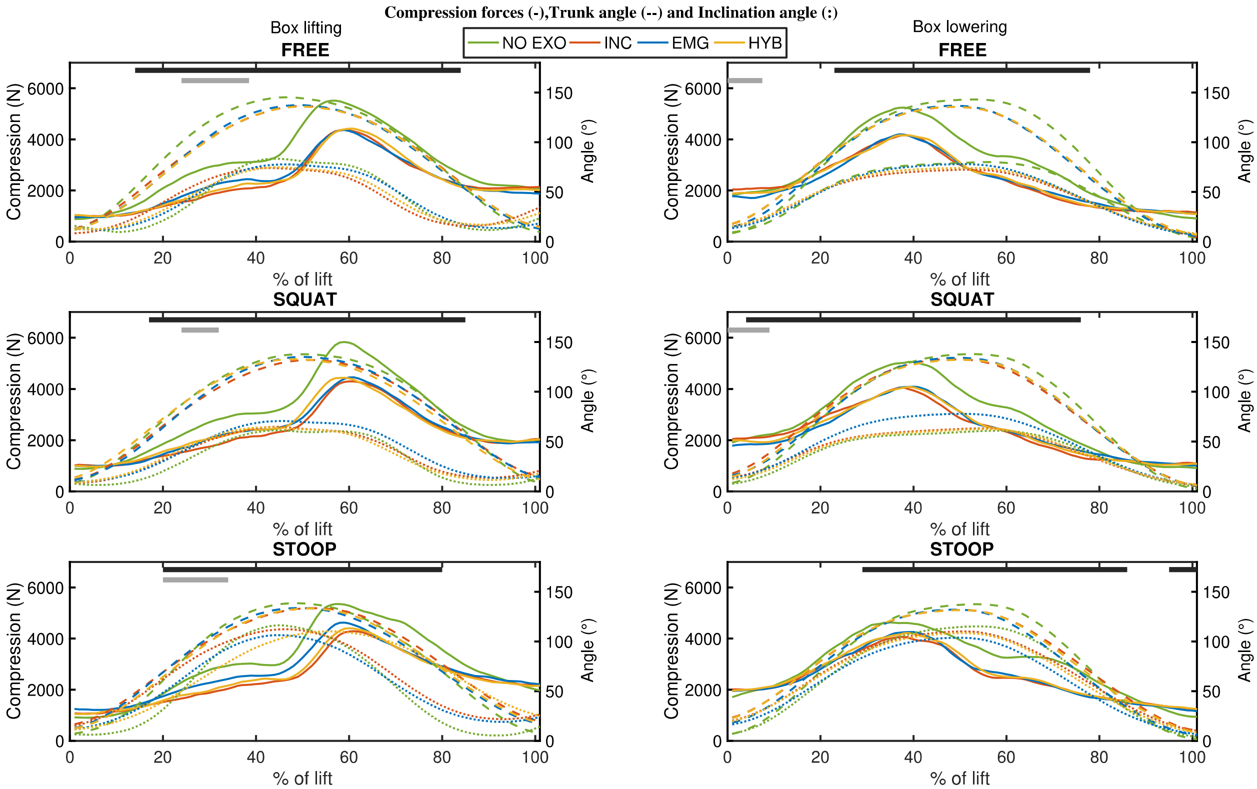

Fig. 5. Time series of compression forces (solid lines), trunk flexion (combined hip and lumbar flexion) angles (dashed lines) and inclination angles (dotted lines) over the whole lifting cycle averaged over participants, separate for box lifting and box lowering. Dark grey horizontal bars indicate the instants that a main effect of control (NO EXO, INC, EMG & HYB) was found. Light grey horizontal bars indicate the instants a main effect of EXO control (INC, EMG & HYB) was found.

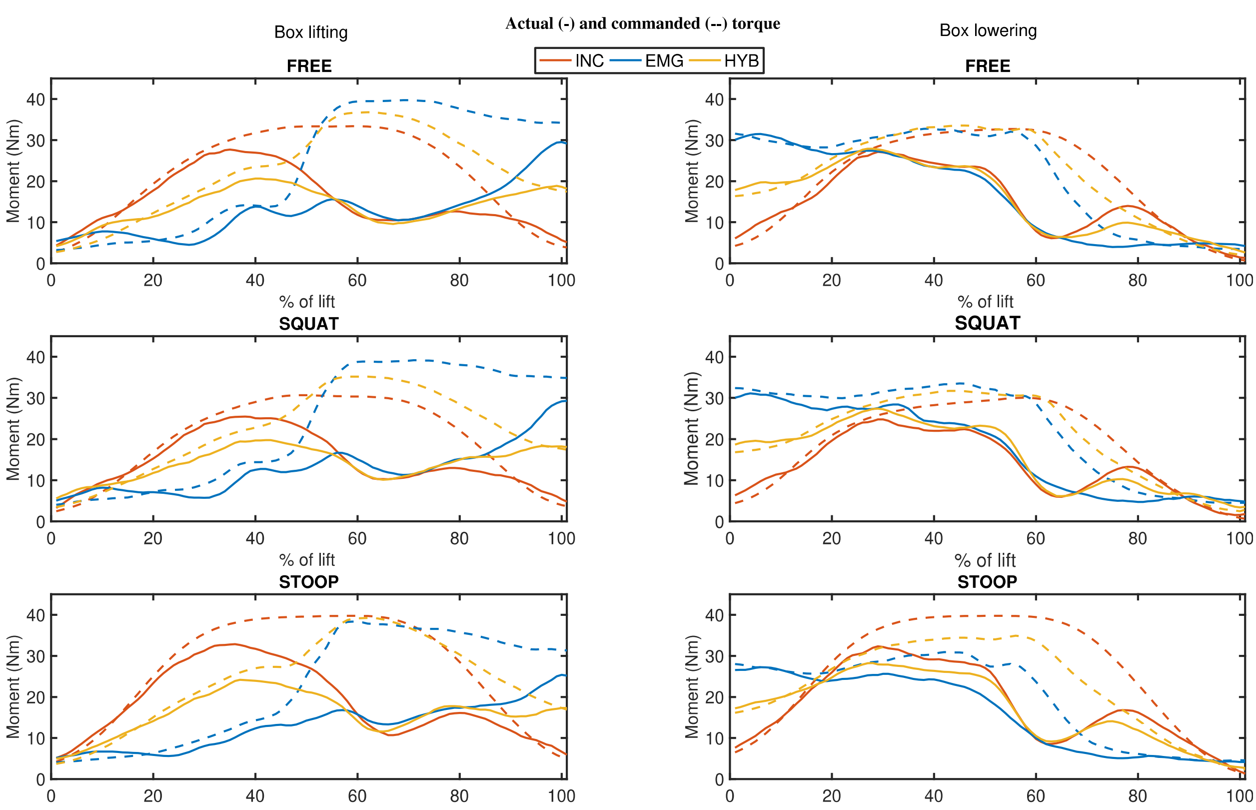

Fig. 6. Time series of the actual applied torques by the EXO (solid lines) and commanded torques (dashed lines) over the whole lifting cycle averaged over participants, separate for box lifting and box lowering.

erations (decelerating the downward movement) and negative angular velocities (upward movement), were much lower than the internal commanded torques (Fig. 6).

4. Discussion

In line with our hypothesis, compression forces with the EXO were substantially lower compared to NO EXO. However, no single EXO control mode was superior over the others due to performance limitations of the actuators. Wearing the active trunk exoskeleton, averaged over all control modes and lifting techniques, reduced peak compression forces by 17.8%. Peak $M_{L5S1\_subject}$ was also substantially reduced by 12.6% compared to NO EXO. These reductions were partly attributable to a reduction of 25% in peak lifting speed when wearing the EXO. Peak back muscle activity was, on average, 13.3% MVC lower compared to NO EXO. Peak lumbar flexion was strongly, by on average 33%, reduced when wearing the EXO. Significant differences were found in the timeseries of the compression forces between the EXO control modes (INCLINATION, EMG & HYBRID), especially during forward bending without box and standing upright with box (Fig. 5).

In spite of the mass of the EXO, which would directly increase low-back load, a substantially lower peak compression force was found when participants used the EXO compared to performing the same task without EXO. Roughly half of the reduction in peak $ML5S1_{subject}$ was attributable to the support of the EXO, the other half due to changes in lifting behavior (25% reduced lifting speed and 33% less lumbar flexion), which reduced peak ML5S1_total by around 15 Nm. The reduction in peak compression force was independent of the instructed lifting technique and the three different control modes, which underscores the robustness of the effect of the EXO. Note however that differences in lumbar flexion between lifting styles became less prominent when wearing the EXO.

The reason why such a strong reduction in lifting speeds occurred with wearing the EXO is not clear. One reason can simply be the added mass of the EXO. It can also be due to friction in the motors at higher speeds, the inertia of the actuators was around 1 kg m^2. However, with the closed-loop control the effective inertia should be six times lower. Another reason might that people were more careful when wearing the EXO in combination with all the markers, although participants did mention that they did not feel limited by it.

The peak compression forces found in the present study (6000 N), are high enough to cause damage to some spines when tested in vitro (Brinckmann et al., 1989). Therefore, the EXO could substantially reduce the population at risk, during manual materials handling. Moreover, compression forces with the EXO were substantially reduced (18.7%) over the whole lifting cycling, so that cumulative loading and the risk at LBP would be substantially reduced as well (Coenen et al., 2013).

Not many studies have investigated the effect of an active exoskeleton on spinal compression forces. But, the approximately 20% reduction in back muscle activity was comparable to Huysamen et al. (2018) who evaluated a variation of the same device with INCLINATION control only. The reduction found in back muscle activity was somewhat lower compared to reductions reported for other active exoskeletons, which ranged from 25% to 60% of back muscle reduction (Kadota et al., 2009; Ko et al., 2018; Kobayashi et al., 2009; Kobayashi and Nozaki, 2008; Li et al., 2013; Muramatsu et al., 2011). This may in part be due to the limited performance of the actuators in the present study. However, to what extent lumbar flexion and lifting speed changed between conditions and how much torque was generated by the EXO remains unclear in these studies.

The fit of the EMG-driven model, used to estimate compression forces, was acceptable with an R^2 between 0.79 and 0.89, which is comparable to other EMG assisted modeling studies (Marras and Granata, 1997; van Dieen and Kingma, 2005). The peak compression forces found in this study are within the range of expected values during dynamical lifting of loads of around 15 kg (Bazrgari et al., 2008; Kingma et al., 2016; Marras and Davis, 1998).

Regrettably, the EXO was not in all cases capable of generating the commanded torques, resulting in the absence of significant differences in compression forces between control modes, which would have been expected based on the commanded torques. In some instants (Fig. 6), the EXO moment was reduced by up to 75% relative to the commanded torque. Therefore, with the current motor limitations it is hard to draw conclusions on what control mode is the most effective.

To predict the potential of the control modes, we therefore analyzed the commanded torque in more detail. In contrast with our hypothesis, the commanded torque did not reach the 40 Nm commanded peak in each mode. In fact, the timeseries of the commanded torque suggest that EMG mode would be most effective during peak loading (10 Nm more compared to INCLINATION). From the different lifting techniques, it can be concluded that the differences in the commanded torque were a result of the EXO inclination angle. During stoop lifting (where inclination is biggest) no differences were observed between control modes, whereas during squat lifting differences were biggest. Apparently, during SQUAT and FREE lifting, 90° EXO inclination was not reached, so that in INCLINATION and HYBRID maximum torque was not achieved.

From a biomechanical perspective, it makes sense to have a control mode supporting the moments required to compensate for forward bending and also compensate for the additional loading required in the case of lifting, the object lifted. Indeed, based on the commanded torque profiles, HYBRID seems to be optimal, as it is effective during forward bending (23 Nm), peak loading (35 Nm) and when standing straight while holding a box (18 Nm). With the successful implementation of the bracelet, HYBRID mode is sensitive to both moment components and unobtrusive to the users. The current setup could be easily implemented in industrial settings, unlike for example control based on trunk muscle activity.

It is beyond the scope of this study to analyze in detail the reasons for the large deviations between the commanded and actual torques. However, it seemed that the deviation started at the instant of deceleration of movement. A second drop in support moment occurred at the instant that the movement changed direction, from bending forward to moving upright. In these situations, a large portion of the torque capability of the actuator is used to accelerate its own rotational inertia, resulting in a substantially reduced output torque.

Some potential sources of bias in this study should be carefully considered. Errors in spinal forces estimated by our EMG-driven model may be due to factors such as cross-talk, bad representation of deep and wide muscles, EMG normalization, ignoring spine translations and considerations of L5S1 moments only (Arjmand et al., 2009; DeLuca and Merletti, 1988; Gagnon et al., 2011; Staudenmann et al., 2005; Stokes et al., 2003). However, these sources of error are not likely to affect our comparison between control modes, as these sources of error are not likely to vary strongly between the control conditions. In addition, we assumed that all mass of the EXO was at the trunk segment. Therefore, we did not separately take this mass into account as it is fully captured in the bottom-up inverse dynamics through the ground reaction force. However, we cannot exclude that a minor portion of the weight of the EXO was transmitted to the pelvis directly. This could not be estimated and was therefore not considered.

In addition, it should be noted that the EXO tested is still in an experimental stage and it isn't a commercially available product. Improvements of the motors and a weight reduction are necessary to make it applicable for industry. Furthermore, subjects reported that they were quickly used to the EXO. However, it cannot be excluded that a longer familiarization period would lead to different results. Besides, the scope of this study was to investigate the effects of the EXO on the low back, in this study we didn't check for additional loading in other joints which should be investigated in future studies. Besides, only male subjects were measured, whereas lifting strategy might differ between males and females (Lindbeck and Kjellberg, 2001; Marras et al., 2002, 2003; Plamondon et al., 2017). Therefore, experiments should be repeated with females.

In conclusion, the EXO reduced compression on the spine by augmenting the moments generated by the trunk extensor muscles and through changes in lifting behavior. This suggests wearing the EXO could reduce the risk of low back pain. With the current limitations of the torque generators it was impossible to determine which of the three control modes was the best. However, based on the commanded torque, we speculate that HYBRID offers a good combination of reducing the mechanical loading associated with bending forward and that associated with lifting loads. Improving the actuators, such that the commanded torque can indeed be generated, would make the EXO more effective in reducing back load.

Declaration of Competing Interest

The authors state that there is no conflict of interest to report.

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